Approximately 70% of the population in industrialized countries experience back pain at least once in the course of their lives. Patients require long-term care, and their quality of life is often limited. Acute episodes of low back pain can be caused by disc prolapses, a multifactorial process in which the mechanical environment as well as age and degeneration effects have an impact. Adams and Hutton, for example, found that high loading can distort the lamellae in the anulus forming radial fissures so that prolapses may occur.2

In the past, many in vivo and in vitro studies were performed to describe the potential overloading of the spine. It was often demonstrated that a disc may prolapse under certain load combinations of flexion, lateral bending, axial rotation, and axial compression.2–7 These combinations may produce high pressures in the nucleus and local regions of large stresses in the anulus,8 with the highest stresses found in the posterolateral region.9 In contrast, an axial rotation alone does not seem to overload the intervertebral disc. This may be due to the facet joints, which transfer a substantial part of the load and thus limit the movement of the disc.10

In experimental in vivo or in vitro investigations, only certain parameters can be measured, e.g., the relative movement between 2 adjacent vertebrae, disc bulges, or the intradiscal pressure in individual areas of the intervertebral disc. Other parameters, such as strains or stresses in different regions of the intervertebral disc, cannot be characterized completely in experiments. In the past, it was shown that finite element (FE) models were helpful to quantify these parameters. However, most of the previous FE studies were only used to simulate pure moments in 1 of the 3 anatomic main planes, simulating flexion-extension, lateral bending, and axial rotation, partially combined with an axial compression. However, in the physiologic situation, a state of complex loading exists. Only few groups analyzed the disc behavior under predefined load combinations.11,12 The investigators examined the mechanical behavior of the disc under specific load combinations, known to result in disc prolapses in vitro: combinations of flexion or extension plus lateral bending and axial rotation. They found that the maximum fiber strains occurred in the posterior and posterolateral anulus and reasoned that disc failure predominantly occur in these areas. However, in these FE studies, only a few load situations were investigated. Other load combinations could lead to higher internal fiber and shear strains in the anulus than those observed combinations. Furthermore, the influence of parameters, such as shear and fiber strains in the anulus on disc failure, has not yet been extensively investigated.
Therefore, the aim of this study was to find load combinations, which would lead to the highest internal stresses in the intervertebral disc and to determine the location in which the highest stresses occurred. To estimate the mechanical behavior of the disc, the shear and fiber strains in the anulus as well as the intradiscal pressure in the nucleus were determined.

**Materials and Methods**

**FE Model.** A nonlinear, 3-dimensional, symmetric FE model of a human lumbar spinal segment L4–L5 was generated based on volume reconstruction of a high-resolution computer tomography scan (Philips MX 8000 IPT device) having a lateral resolution of 0.49 mm with a slice thickness of 0.75 mm (Figure 1). Additional magnetic resonance imaging (Magnetom Symphony Maestro Class, Siemens, Germany) and histologic observations were conducted defining the soft tissue geometries. The reconstructed volume data set was transferred into a FE package (ANSYS 10.0; Swanson Analysis, Houston, PA) and subsequently meshed. The modeled vertebrae included cortical bone, cancellous bone, bony endplates, and posterior structures with facet joints. These components and the intervening intervertebral disc with the cartilaginous endplate were meshed using 8-node isoparametric solid elements. The collagen fibers of the anulus and the 7 spinal ligaments, the anterior and posterior longitudinal ligament, flaval, supraspinous, interspinous, transversal, and capsular ligaments were represented by 3-dimensional, unidirectional spring elements. The contact between the facet joints was simulated by surface-to-surface contact elements without friction.

The modeled intervertebral disc consisted of the nucleus pulposus and the anulus, whereas the anulus was assumed to be composed of a homogeneous ground substance reinforced by a collagen fiber network (Figure 1). Eight crisscross fiber layers were defined in radial direction. The angulations of the fibers varied from $\pm 24^\circ$ to the horizontal plane ventrally to $\pm 46^\circ$ at the dorsal side according to histologic findings. The relative volume content of the fibers with respect to the surrounding ground substance was assumed to vary from 23% at the outer layer to 5% at the inner fiber layer.

Material properties of the different tissues were extracted from the literature (Table 1). The fluid-like behavior of the nucleus and the hyperelastic properties of the anulus ground substance were both modeled using an isotropic, incompressible, hyperelastic Mooney-Rivlin formulation. The stress-strain behavior of the anular collagen fibers were described by a nonlinear function, which was obtained from previous reports. Since outer lamellae behave stiffer than inner lamellae, the fibers in different anulus layers were weighted (outermost layers 1–2, 1.0; layers 3–4, 0.9; layers 5–6, 0.75; Table 1. Material Properties of the Different Tissues in the Finite Element Model

<table>
<thead>
<tr>
<th>Material</th>
<th>Young’s Moduli (MPa)</th>
<th>Poisson’s Ratio</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical bone</td>
<td>E\textsubscript{xx} = 11,300</td>
<td>$\nu_{xy}$ = 0.484</td>
<td>Lu \textit{et al}\textsuperscript{11}</td>
</tr>
<tr>
<td></td>
<td>E\textsubscript{yy} = 11,300</td>
<td>$\nu_{yz}$ = 0.203</td>
<td></td>
</tr>
<tr>
<td></td>
<td>E\textsubscript{zz} = 22,000</td>
<td>$\nu_{xz}$ = 0.203</td>
<td></td>
</tr>
<tr>
<td></td>
<td>G\textsubscript{xy} = 3800</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>G\textsubscript{yz} = 5400</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cancellous bone</td>
<td>E\textsubscript{xx} = 140</td>
<td>$\nu_{xy}$ = 0.450</td>
<td>Lu \textit{et al}\textsuperscript{11}</td>
</tr>
<tr>
<td></td>
<td>E\textsubscript{yy} = 140</td>
<td>$\nu_{yz}$ = 0.315</td>
<td></td>
</tr>
<tr>
<td></td>
<td>E\textsubscript{zz} = 200</td>
<td>$\nu_{xz}$ = 0.315</td>
<td></td>
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<tr>
<td></td>
<td>G\textsubscript{xy} = 48.3</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>G\textsubscript{yz} = 48.3</td>
<td></td>
<td></td>
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<tr>
<td></td>
<td>G\textsubscript{xz} = 48.3</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Posterior bony elements</td>
<td>E = 3500</td>
<td>$\nu = 0.25$</td>
<td>Shirazi-Adl \textit{et al}\textsuperscript{46}</td>
</tr>
<tr>
<td>Bony endplate</td>
<td>E = 4000 to 12,000</td>
<td>$\nu = 0.3$</td>
<td>Edwards \textit{et al}\textsuperscript{46}</td>
</tr>
<tr>
<td>Cartilaginous endplate</td>
<td>E = 23.8</td>
<td>$\nu = 0.4$</td>
<td>Lu \textit{et al}\textsuperscript{11}</td>
</tr>
<tr>
<td>Anulus ground substance</td>
<td>Mooney-Rivlin $c_1 = 0.18, c_2 = 0.045$</td>
<td>$\nu = 0.45$</td>
<td>Schmidt \textit{et al}\textsuperscript{50}</td>
</tr>
<tr>
<td>Anulus fibers</td>
<td>—</td>
<td></td>
<td>Stress-strain curve determined by Shirazi-Adl \textit{et al}\textsuperscript{46}</td>
</tr>
<tr>
<td>Nucleus pulposus</td>
<td>Mooney-Rivlin $c_1 = 0.12, c_2 = 0.03$</td>
<td>$\nu = 0.4999$</td>
<td>Smit\textsuperscript{51}</td>
</tr>
</tbody>
</table>

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innermost layers 7–8, 0.65). Force-deflection curves were obtained to represent spinal ligament behavior. The facet cartilage was assumed to be multilinear elastic in compression.

Calibration. For the calibration process, in vitro data from previously reported studies were used, including range of motion (rotation) and intradiscal pressure. In these experiments, 6 specimens were tested in the intact state. Afterwards, the anatomy was successively reduced, including the different ligaments, facet joints, and nucleus. In the intact and in all defect stages, the segment was tested with pure moments of 1, 2.5, 5, 7.5, and 10 Nm in all load directions. Before the experiments, specimens were exposed for 15 minutes to 500 N axial compression to reduce the water content of specimens to avoid abnormal height water content.

The FE model, whose geometry was based on 1 of the 6 tested specimens calibrated with these data by adding these structures in the opposite way, starting with an isolated anulus to which only the vertebral bodies were added. The other different anatomic structures were cumulatively added. In each calibration step, the material property of the added structure was modified, so that the FE model fulfilled the experimentally obtained range of motion and intradiscal pressure data.  

Anulus. In a previous study, a calibration method was developed, which considers the individual contribution of the fibers and the ground substance. The stiffness of the fibers was varied to approximate the Young's modulus of the ground substance in order to fulfill the required range of motion.

Nucleus Pulposus. A parametric study was performed on the nucleus material properties. Young's modulus was varied in a range of 0.1 to 4 MPa. Vertebral Arches With Facet Joints. The orientation of the facet joints was varied in a parametric study in order to obtain the influence on the motion response. The final angle is within the reported range.

Ligaments. The force-deflection behavior of all ligaments was sequentially computed. Because of the 5 moment magnitudes, 6 points were determined and subsequently interconnected by a spline function describing continuous force deflection behavior.

Validation. For this study, the results of the complete assembled FE model were additionally compared with intradiscal pressure data from previously performed experimental studies.

Loading and Boundary Conditions. The inferior endplate of the lower vertebral body was rigidly fixed. Pure unconstrained moments of 7.5 Nm were applied to the superior endplate of the upper vertebral body, as it has been recommended. The loading direction was incrementally changed by an angle of 15° between the different anatomic planes to realize not only pure moments in flexion/extension, lateral bending and axial rotation but also load combinations of flexion plus lateral bending, flexion plus axial rotation, extension plus lateral bending, extension plus axial rotation, and lateral bending plus axial rotation. The line of action for the resulting moment between 2 anatomic planes was an oblique spatial axis. The applied moment about this axis was always 7.5 Nm.

Subsequently, these load scenarios were additionally combined with an axial compression of 500 N simulating upper body weight. This load was applied as if it was a follower load. Thus, the load path passed the center of the vertebral bodies and did not additionally create any significant bending moment. For the FE model, nonlinear large deformations were used for calculation. To ensure the convergence, 6 to 10 substeps were iteratively determined using the “Newton-Raphson” approach.

Data Analysis. The following parameters were considered to be the most important to estimate internal stress behavior of the intervertebral disc:

The intradiscal pressure in the nucleus was determined as one third of the trace of the stress tensor, \( t.e., \) the mean of the 3 normal stresses. This was necessary since the nucleus was generated with solid elements.

The shear strains between the anulus and the adjacent endplates were determined as a vector summation of the shear strain components \( e_{xx} \) and \( e_{yy} \). Thereby, \( x \) was defined in posterioranterior, \( y \) in lateral, and \( z \) in longitudinal direction. It was found in radiographic studies that the outer anulus separate from the adjacent vertebral bodies and produce peripheral rim lesions. We assumed that these failures mainly caused by a resulting shear load.

The tensile strains in the normal direction of the fibers, which may lead to fiber disruptions and initiate radial tears.

Results

Validation

Both, the numerical and the in vitro curves represent a similar nonlinear curve (Figure 2). Under a moment of 7.5 Nm, flexion showed with 5.9° the largest range of motion (in vitro, 6.1°), followed by lateral bending with 5.3° (in vitro, 5.15°), extension with 4.5° (in vitro, 4.1°), and axial rotation with 2.5° (2.7°).

Intradiscal Pressure

The intradiscal pressure was highest in flexion (0.35 MPa), followed by extension (0.18 MPa), axial rotation (0.16 MPa), and lateral bending (0.14 MPa) (Figure 3). A bending or torsion moment about an oblique axis did not strongly increase the nucleus pressure compared with the same moment applied in a principle direction. For all load scenarios, the additionally applied follower load resulted in an increase by an average offset value of 0.34 MPa.

Shear Strains

For all load scenarios, the maximum shear strain was found to be located between the anulus and the inferior endplate. Under pure moments, lateral bending generated the largest shear strains (39.7%), while axial rotation (32.8%) led to the smallest shear strains (Figure 4). The maximum shear strains for lateral bending occurred at the ipsilateral side of the anulus (Figure 5).

A combination of lateral bending plus extension and lateral bending plus flexion produced a substantial increase in shear strains (up to 44.7%) (Figure 4). They occurred at the posterolateral region of the anulus (Figure 6). In contrast, a combination of right axial rotation
plus lateral bending and axial rotation plus extension showed a large decrease in shear strains and occurred at the lateral and posterolateral region of the annulus, respectively. The presence of a follower load tended to increase the shear strains for all load scenarios by an average offset value of 5% but did not change the location of the maximum shear strain.

Fiber Strains
Except axial rotation, the maximum fiber strains increased toward the innermost fiber layer (Figures 7 and 8). Under pure moments, it was observed that axial rotation generated the largest tensile strains in the collagen fibers (11.9%), while extension led to the smallest fiber strains (5.9%). Axial rotation showed maximum tensile strains posterolaterally. Only the fibers, which were oriented in direction of the applied moment, underwent tensile strains.

Lateral bending in combination with left axial rotation yielded the highest increase in fiber strains (19.8%) (Figure 7). A load combination of lateral bending plus extension showed the strongest decrease in fiber strains (down to 1.5%). The maximum strains for lateral bending plus axial rotation occurred at the posterolateral region (Figure 8). For all load combinations, the fibers, which run from the inferior endplate to the superior endplate in a clockwise direction, underwent tensile strains.
The fibers running in the other direction were all in compression.

The additionally applied follower load resulted in an increase in the maximum fiber strains for all load scenarios, by an average offset value of 3.3% but did also not change the location of the maximum fiber strain.

**Discussion**

In the present study, a 3-dimensional, nonlinear FE model was used to determine the load combinations, which led to the highest internal loads of the intervertebral disc. The results of this study yielded some general rules, which might be important in clarifying the cause of anulus failure and disc prolapses.

**Validation**

The predicted relationship between range of motion and intradiscal pressure generated by the complete assembled FE model showed a good agreement with the experimentally determined *in vitro* data (Figure 2).

Currently, there is a paucity of *in vitro* data concerning both fiber and shear strain measurements, due to the difficulty in obtaining these measurements without damaging or destroying the intact discs. Therefore, it was not possible to directly validate the fiber and shear strains of the anulus in the FE model. However, to ensure the accuracy of the FE model, the disc behavior was compared with measurements of Shah et al, who determined circumferential strains at the anulus surface in flexion and extension. They reported that the tangential surface strain was highest at the posterior disc for flexion and at the anterior disc for extension. Since the fibers in our model reflect this behavior quite well, we concluded that the fiber strains were in a conceivable range. Tencer and Mayer computed in an experimental kinematical approach that the maximum strains for lateral bending occurred at the contralateral side of the anulus, which also was in good agreement with the presented results. However, a comparison of the fiber strains in axial rotation indicated a disagreement.

Unfortunately, we could not find any literature to validate our shear strains. Since the FE model showed a good agreement with the range of motion, we concluded that the shear strains were in a conceivable range. However, these findings should be interpreted with care.

**Intradiscal Pressure**

An overload of the intervertebral disc during combined loading considering only the intradiscal pressure does not seem to be given. Furthermore, the intradiscal pressure seems to be dependent on the range of motion. Both were highest in flexion and smallest in lateral bending and did not show a maximum under combined load scenarios. Previously performed *in vitro* experiments showed a large range of intradiscal pressures. Yet the results of the presented study showed similar tendencies, especially compared with *in vitro* pressure measurements of McNally et al, who investigated the pressure distribution in the intervertebral disc under a compressive load of 500 N. Similar to the presented study, the authors found that the intradiscal pressure increased by 0.5 MPa.

**Fiber Strains**

It was shown that, under pure moments, axial rotation generated the largest tensile strains in the fibers. It seems to be contrary to a previous *in vitro* study. There it was stated that the intervertebral disc is protected by the apophyseal joints against axial rotation. However, the fiber strains would not substantially change when higher moments in axial rotation are applied (7.5 Nm, 11.9%; 10 Nm, 12.2%), while moments in flexion (7.5 Nm, 7.2%; 10 Nm, 9.2%), extension (7.5 Nm, 5.9%; 10 Nm, 8.3%) and lateral bending (7.5 Nm, 8.9%; 10 Nm, 12.8%) strongly increase the fiber strains. In axial rotation, from 7.5 Nm upwards, the fibers are protected by the facet joints. This correlates with a previous numerical study by...
Shirazi-Adl and approves in vitro findings of Adams and Hutton. The fibers underwent a maximum tensile strain during load combinations of axial rotation with lateral bending and axial rotation with flexion. These load conditions essentially affect the innermost annulus layer at the posterolateral location. This was comparable with previously reported FE studies. They suggested that lifting combined with bending and axial rotation could be responsible for initiating fiber failure at the inner anulus layer in the posterior and posterolateral region. During an optimization study, it was found that anterior fibers need to be 32% stiffer than posterolateral fibers to fulfill the in vitro flexibility of the anulus. These results may explain the high level of tensile strains and therefore eventual ruptures of fiber in this region. Under pure moments, axial rotation generated the largest fiber strains, whereas extension resulted in the smallest strains. It should be noted that the range of motion at the lumbar segment was higher in extension than in axial rotation.

The strains in the anulus fibers increased essentially when an additional follower load was applied. In experimental studies, it was found that the ultimate tensile strain of collagenous fibers is 10% to 25%. This suggests that, under lateral bending plus left axial rotation and axial rotation plus flexion even without a follower load, disc fibers in the posterolateral region may be susceptible to rupture.

**Shear Strains**

It was found that the load combinations, which caused a strong increase of the fiber strains, did not also lead to the largest increase of the shear strains. The anulus underwent a maximum shear strain exposed to lateral bending plus flexion or extension. In comparison to the fiber strains, the maximum shear strains occurred also posterolaterally, which is consistent with previous findings. Furthermore, the maximum shear strains were located caudally close to the endplate. This suggests that tears could occur at the interface to the lower rather than to the upper endplate. Thus, according to these results, a disc prolapse could be located posterolaterally at the inferior endplate, which would be consistent with previous reports.

**Limitations of the FE Analysis**

During the segmentation and reconstruction process, the geometries of both vertebrae were smoothed to limit the number of elements. More anatomic details would re-
quire substantially more elements and nodes, including more degrees of freedom. These additions would have dramatically increased the computation time. The geometry of the anulus was based on transverse histologic slices of specimens and magnetic resonance imaging scans. However, resolution and slice thickness of both methods were limited.

A variation in geometric parameters, such as disc height, cross-sectional area of the intervertebral disc, size and position of the nucleus, fiber network orientation, or the number of fiber layers, can affect the mechanical behavior of the intervertebral disc.24,47–49 This suggests that other FE models with different geometries might lead to different results. However, after careful validation, all FE models should show at least the same tendencies.22

**Conclusion**

The study showed that the anulus is highest strained in the posterolateral region. This might explain that the most common location of lumbar disc prolapse occur in this location. A disc may prolapse under a combination of axial rotation plus lateral bending, axial rotation plus flexion or lateral bending plus flexion or extension. This risk will be significantly increased when an axial load is also added. For clinical practice, this would mean that patients should avoid load combinations, especially with lifting tasks.

**Key Points**

- The aim was to find the load combination, which leads to highest pressure in the nucleus and shear and fiber strains in the anulus.
- A 3-dimensional, calibrated finite-element model of a functional spinal unit (L4–L5) was used.

- Intradiscal pressure correlated with the magnitude of deformation.
- The maximum shear and fiber strains occurred during combined bending moments and were located posterolaterally.
- Results may help clinicians to better explain the mechanical cause of disc failure and disc prolapses.

**Acknowledgments**

The authors thank Dr. B. Willie for editorial assistance.

**References**
